Mechanics of the Knee

A study of joint and muscle load with clinical applications

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To Susanne
To my parents

ABSTRACT

The load moment of force about the knee joint during machine milking and when lifting a 12.8 kg box was quantified using a computerized static sagittal plane body model. Surface electromyography of quadriceps and hamstrings muscles was normalized and expressed as a percentage of an isometric maximum voluntary test contraction. Working with straight knees and the trunk flexed forwards induced extending knee load moments of maximum 55 Nm. Lifting the box with flexed knees gave flexing moments of 50 Nm at the beginning of the lift, irrespective of whether the burden was between or in front of the feet. During machine milking, a level difference between operator and cow of 0.70 m – 1.0 m significantly lowered the knee extending moments.

To quantify the force magnitudes acting in the tibio-femoral and patello-femoral joints, a local biomechanical model of the knee was developed using a combination of cadaver knee dissections and lateral knee radiographs of healthy subjects. The moment arm of the knee extensor was significantly shorter for women than for men, which resulted in higher knee joint forces in women if the same moment was produced. A diagram for quantifying patellar forces was worked out. The force magnitudes given by the knee joint biomechanical model correlated well with experimentally forces measured by others.

During the parallel squat in powerlifting, the maximum flexing knee load moment was estimated to 335 – 550 Nm when carrying a 382.5 kg burden and the *in vivo* force of a complete quadriceps tendon-muscle rupture to between 10,900 and 18,300 N. During isokinetic knee extension, the tibio-femoral compressive force reached peak magnitudes of 9 times body weight and the anteroposterior shear force was close to 1 body weight at knee angles straighter than 60 degrees, indicating that high forces stress the anterior cruciate ligament. A proximal resistance pad position decreased the shear force considerably, and this position is recommended in early rehabilitation after anterior cruciate ligament repairs or reconstructions.

The methods presented quantify muscle activity, sagittal knee joint moments and forces, enabling assessments to be made of different work postures, training exercises and joint derangements.

This thesis is based on the following papers which will be referred to in the text by their Roman numerals I-VI.

I. Ekholm J, R Nisell, UP Arborelius, C Hammarberg & G Németh: Load on knee joint structures and muscular activity during lifting. Scandinavian Journal of Rehabilitation Medicine, 16:1–9, 1984

II. Ekholm J, R Nisell, UP Arborelius, O Svensson & G Németh: Load on knee joint and knee muscular activity during machine milking. Ergonomics, 28:665–682, 1985

III. Nisell R, G Németh & H Ohlsén: Joint forces in extension of the knee. Analysis of a mechanical model. Acta Orthopaedica Scandinavica, Accepted for publication, 1985

IV. Nisell R & J Ekholm: Patellar forces during knee extension. Scandinavian Journal of Rehabilitation Medicine, 17:63–74, 1985

V. Nisell R & J Ekholm: Joint load during the parallel squat in powerlifting and force analysis of *in vivo* bilateral quadriceps tendon rupture. Submitted for publication, 1985

VI. Nisell R, MO Ericson, G Németh & J Ekholm: Tibio-femoral joint forces during isokinetic knee extension. Submitted for publication, 1985

CONTENTS

Abbreviations and definitions

BW	= Body weight
C	= Centre of contact pressure point between femoral and tibial condyles
deg	= Degree
de	= Moment arm of external force acting on tibia to bilateral knee axis
dp	= Knee extensor moment arm, perpendicular distance of patellar tendon force to tibio-femoral contact point
EMG	= Electromyography
Fcp	= Compressive force in patello-femoral joint, perpendicular to joint surface
Fcq	= Compressive force between quadriceps tendon and femoral intercondylar groove
Fct	= Compressive force in tibio-femoral joint, perpendicular to joint surface
Fet	= External force acting on tibia tending to flex the knee
FKC	= Lift with flexed knees and box between feet (close to body)
FKFF	= Lift with flexed knees and box in front of feet (far from body)
Fp	= Force in patellar (infrapatellar) tendon
Fq	= Force in quadriceps (suprapatellar) tendon
Fqm	= Force in quadriceps muscle
Fs	= Anteroposterior shear force in tibio-femoral joint along joint surface. Positive value indicates when tibia tends to slide anteriorly in relation to femur, negative when sliding posteriorly
kg	= Kilogram
m	= Meter
N	= Newton
Nm	= Newtonmeter
sec	= Second
SK	= Lift with straight knees
TAMP-R	= Time averaged myoelectrical potential ratio, *i.e.* muscle activity recorded during the experiment divided by the reference level recorded during an isometric maximum voluntary test contraction (normalized EMG)
WCR	= Working cycle ratio during lifting, runs from 0 to 1

α (alfa)	= Knee flexion angle, between long axes of tibia and femur. Straight knee equals 0 deg
β (beta)	= Angle between Fp and perpendicular line of tibial plateau. Positive when tibia is pulled anteriorly in relation to femur
γ (gamma)	= Angle between Fp and long axis of tibia. Positive when the intersection is below the tibial plateau
δ (delta)	= Angle between external force and tibial plateau
ε (epsilon)	= Angle between Fq and Fcp
λ (lambda)	= Angle between Fq and Fqm
ψ (psi)	= Angle between Fp and Fq
ω(omega)	= Angle between tibial plateau and perpendicular line of tibial long axis

Introduction

The concept of biomechanics may be defined in different ways. A definition recently suggested by LeVeau (1984) is: "the study of forces and the effects of those forces on and within the human body". Hence, musculoskeletal load in all its aspects becomes a substantial part of the biomechanical concept. Biomechanics has become an interdiscipline between existing fields such as mechanical engineering which develop methods, equipment and accuracy, and medicine where for example ergonomics, orthopaedics and rehabilitation abound with practical questions and applications. As a consequence, one of the main goals for the medically and biomechanically trained, might be to understand and use engineering knowledge and methods to solve practical, clinical and preventive problems seen in medical care among patients. In this way the biomechanical field of science may function as a bridge between medical and engineering specializations and, as such, it has become a new and most fascinating field with a variety of practical applications (*e.g.* Frankel & Burstein 1972, Carlsöö 1975, LeVeau, Williams & Lissner 1977, Winter 1979, Frankel & Nordin 1980, Dowson & Wright 1981, Maquet 1984, Chaffin & Andersson 1984).

Quantifying joint moments, forces and muscle activities in various situations or during different activities may be of value from the medical viewpoint, with applications in three different areas:

1) In *ergonomics*, to study the amount of mechanical load and thus to indicate possible load reduction during work; to assess harmful effects of work situations and for designing work equipment; to reduce pain from the locomotor system and load elicited diseases; to improve possibilities of assessing the long-term cause of workload-related diseases and, in a longer perspective, to reduce the number of sick-leave days due to such symptoms.

2) In *orthopaedic surgery*, to assess mechanically the effects of bone and joint derangements (such as those following osteotomy, patellectomy or fractures) and thus to influence the pre- and postoperative investigation of patients; to obtain criteria for material strength in the design of endoprosthesis; to optimize post-surgical rehabilitation after for example anterior cruciate ligament repairs and reconstructions.

3) In *rehabilitation*, to individualize and optimize physical therapy without risks for damaged, injured, diseased or weak structures; to favour exercises that entail a high degree of muscle activity while keeping the load on joint structures low; to estimate the physiological load during normal activities (*e.g.* walking, running, stair climbing, taking a step down, rising from a chair) so as to avoid extreme loads; to improve movement patterns and performance and avoid harmful exercises, which is important also in sports medicine.

"Joint load" is not a very well-defined concept (Németh 1984). The moment of force (= moment, sometimes also called torque) acting about a joint axis is often a useful way to express either load (= load moment) or strength (= muscle moment). However, for different joint angles but the same moments, the joint forces may vary considerably due to changes in muscle moment arm lengths and, consequently, the muscle force magnitudes change. Quantifying the joint force is preferred as it better describes the loading effect on the joint surface. The joint force can be either in the normal (= compressive) or tangential (= shearing) direction of the joint surface or, most commonly, a combination of both. The joint compressive force acts over an area, which indicates that another way to express joint load would be in terms of pressure (N/m^2). This is probably a very good parameter of "joint load". When the menisci are removed from the knee joint, the compressive force magnitude will remain the same, but the pressure will be larger as the tibio-femoral contact area is considerably smaller (Walker & Erkman 1975, Seedhom 1979, Fukubayashi & Kurosawa 1980). This increased joint load is probably the main reason why degenerative changes are frequent many years after meniscectomy (*e.g.* Fairbank 1948, Jackson 1968, Johnson et al. 1974). In this thesis the expression "knee joint load" is synonymous with knee load moment and/or joint force.

"Muscle load" can be studied in many ways. Direct EMG was earlier used frequently (*e.g.* Basmajian 1962, Pocock 1963, Jonsson & Synnerstad 1966, Hallén & Lindahl 1967, Carlsöö 1975). Today normalized and thus quantified EMG has been developed and has definite advantages as it is possible to compare the EMG-level between muscles of different subjects (Ekholm et al. 1978, Philips & Petrofsky 1983, Reynolds et al. 1983, Soderberg & Cook 1983, Andriacchi et al. 1984, Brask et al. 1984, Németh 1984, Yang & Winter 1984). Another approach is to quantify muscle load as a ratio between the muscle moment produced during an activity and the maximum strength measured during an isometric voluntary test contraction (Schüldt et al. 1983, Németh et al. 1984, Svensson et al. 1985). These kinds of "muscle load" quantification were used in Studies I and II where the EMG was normalized and expressed as a percentage of an isometric maximum voluntary test contraction. Muscle load has also been frequently studied by EMG frequency or amplitude distribution, which makes fatigue estimations possible (*e.g.* Kadefors et al. 1968, Komi & Tesch 1979, Kadefors 1978, Petrofsky 1979, Hagberg 1979, Herberts et al. 1980). Muscle force magnitude as a function of EMG is not always linear, especially at muscle activities close to maximum (*e.g.* Möller 1966, Zuniga et al. 1970, Bouisset 1973, Vredenbregt & Rau 1973, Jonsson 1978, Woods & Bigland-Ritchie 1983). The EMG-force relation seems to change between muscles in the same individual and between the same muscle of different individuals, probably due to differences in muscle fibre composition (Woods & Bigland-Ritchie 1983). So, unfortunately, the muscle force magnitudes exerted cannot easily be read from the EMG.

There are, in principle, two different methodological approaches of quantifying joint load; 1) *experimental measurements* and, 2) *biomechanical models* and of course these two methods may be combined. In experimental measurements, force, pressure, etc., are measured directly by a specific sensor (*e.g.* strain gauge, force plate, pressure sensitive film). In the biomechanical modelling approach, the unknown parameter is calculated indirectly using reasonable approximations, mechanical laws and if any, only a few pure experimental measurements. The advantage of a model, if good, is that, besides being non-invasive, it can be generally applied to many different situations without performing new, time-consuming experiments. However, it is important that such a model is realistic. Its validity can be studied by comparing experimental measurement results with results from the model. If there is a bad correlation, either the experimental set up or the biomechanical model solution, or both, will be inadequate for describing the reality.

This thesis presents biomechanical methods for quantifying knee joint load during various activities and exercises. Examples of knee load magnitudes are reported and the effects and implications of the induced load are discussed. The present work is one part of a series of investigations from the Kinesiology Research Group (*e.g.* Ekholm et al. 1978, Arborelius & Ekholm 1978, Ekholm et al. 1982, Schüldt et al. 1983, Harms-Ringdahl et al. 1983, Németh 1984, Svensson et al. 1985, Ericson et al. 1985a and 1985b) concerning levels of muscle activity and the loads on the major joints during various activities and exercises.

Aims of the Study

The general aim of the present investigation was to quantify the load induced to the knee joint in some specific situations or activities. At the beginning of the work on this thesis, only the knee load moment was quantified (Studies I and II), but it was soon obvious that there was a need for a local knee biomechanical model to determine the joint forces induced during the activities. Therefore, a general knee model was worked out (Studies III and IV). The purpose of this model was to be a simplistic and useful, but still realistic, two-dimensional biomechanical model that could be applied to any knee extending activity. In Studies V and VI, examples of how the model can be applied are given.

More specifically, the following topics have been studied:

(1) The muscle activity and load moment about the bilateral knee axis during different standing work postures with the trunk flexed forwards (machine-milking) and during lifting a box, with the knees either flexed or straight.

(2) A local knee biomechanical model for quantifying the forces in the tibio-femoral joint and in the patello-femoral joint.

(3) The *in vivo* force magnitude in the quadriceps tendon at the moment of a traumatic rupture.

(4) The magnitude of the tibio-femoral joint forces during isokinetic knee extension and the stress on the anterior cruciate ligament.

(5) The clinical and practical applications of the results in fields such as ergonomics, orthopaedics and rehabilitation

Literature review

Knee load in different work postures

When studying the mechanical load during work with the trunk leaning forwards, the main interest hitherto has been focused on the load induced to the low back (*e.g.* Andersson et al. 1976, Ekholm et al. 1982, Nordin 1982, Chaffin & Andersson 1984). However, dysfunction of the knee joint is also, along with that of the other major joints, a common cause of impaired working capacity among unskilled labourers (Undeutsch et al. 1982). Mechanical knee joint loads in different standing work situations have not earlier been reported in the literature.

Tibio-femoral joint

In a biomechanical model of knee extension, the patellar tendon moment arm length is decisive for the magnitudes of the forces calculated. The moment arm length has been determined mainly in three different ways;

1. Haxton (1945) and Kaufer (1971) calculated the moment arm at different knee flexion angles by using measured data of flexing knee moments which were related to counteracting forces from the knee extensor. The same method was also recently used by Wendt & Johnson (1985).

2. Lindahl & Movin (1967) and Lindahl et al. (1969) using lateral radiographs showed how the tibio-femoral force magnitudes may be calculated when using the distance between the tibio-femoral contact point (C) and the patellar tendon as the knee extensor moment arm. Several authors have subsequently used a similar "tibio-

femoral contact" approach in their biomechanical knee models (*e.g.* Reilly & Martens 1972, Haffajee et al. 1972, Perry et al. 1975, Seedhom & Terayama 1976, Bishop 1977, Ellis et al. 1979).

3. The moment arm length has been determined on lateral x-rays as the perpendicular distance between the sagittal knee centre axis of rotation and the patellar tendon (Bandi 1972, Smidt 1973). Smidt (1973) presented useful average data of knee joint moment arms from 26 normal men in order to calculate the joint forces during maximum isometric knee extension and flexion.

Bresler & Frankel (1950), using ground reaction forces and mechanical equations, calculated the maximum extending muscular moment about the knee joint during human gait to between 40 Nm and 88 Nm. Similar methods have subsequently been used for studying joint load during level walking (Radcliffe 1962, Winter 1980, Boccardi et al. 1981) and by Morrison (1968, 1969 and 1970) who related the calculated load to anatomical data of one knee dissection in order to determine knee joint and ligament forces. Morrison (1970) calculated the maximum mean tibio-femoral compressive force (Fct) to about 3 times BW during level walking. When walking up stairs the maximum Fct was above 4 BW and down stairs just below 4 BW (Morrison 1969). Andriacchi et al. (1980) calculated the knee extensor muscular moment to 54 Nm when ascending stairs and as high as 147 Nm when descending.

Tibio-femoral joint load has also been quantified by means of biomechanical models in other activities such as flexed knee stance and rising from seated and squatting positions (Seedhom & Terayama 1976, Bishop 1977, Ellis et al. 1979 and 1984, Kelley et al. 1978, Dahlqvist et al. 1982, Schüldt et al. 1983), jumping (Alexander & Vernon 1975, Smith 1975, Nisell et al. 1985), kicking (Wahrenberg et al. 1978, Lindbeck 1983) and bicycling (Gregor et al. 1985, Ericson et al. 1985b).

Patello-femoral joint

Several biomechanical models have suggested how to determine the magnitude of the patello-femoral compressive force (Fcp). The oldest and most easily-used model assumes the quadriceps (Fq) and patellar (Fp) tendon forces to have the same directions as the tibial and femoral long axes. Other, more advanced, models may consider the angle of the patellar tendon in relation to the tibial long axis (Reilly & Martens 1972, Bandi 1972, Smidt 1973, Ellis et al. 1979, Hungerford & Lennox 1983) or as Matthews et al. (1977), who suggested a special linear relationship between the knee flexion angle and angle ψ. However, all these models consider Fp to have the same magnitude as Fq throughout the whole range of motion.

As early as 1945, Haxton showed that the patellar and quadriceps tendon forces for the flexed knee joint are not of the same magnitude, the patellar tendon taking up lower forces than the quadriceps tendon. In recent years several authors have reported the same at knee flexion angles above 60 deg, where the Fp/Fq ratio is suggested to range between 0.50 – 0.75 (Bishop 1977, Bishop & Denham 1977, Ellis et al. 1980, Maquet 1984, Huberti et al. 1984).

Reilly & Martens (1972) calculated Fcp to 0.5 times BW during level walking and 3.3 BW during stair-climbing. Smidt (1973) calculated Fcp at isometric maximum knee extension to 2.6 BW. Dahlqvist et al. (1982) estimated maximum Fcp values during squatting to between 4600 N and 5900 N or about 7 BW. Fcp has also been estimated during rising from a chair with and without the aid of the arms (Seedhom & Terayama 1976, Ellis et al. 1979 and 1984).

Quadriceps rupture

Bilateral simultaneous complete rupture of the quadriceps tendons is a rare injury with 24 cases reported in the literature (Frey 1928, Goldenberg & Paterson 1949, Steiner & Palmer 1949, Wetzler & Merkow 1950, Wilson 1957, Preston & Adicoff 1962, Dalal & Whittam 1966, MacDonald 1966, Levy et al. 1971, Firooznia et al. 1973, Brotherton & Ball 1975, Peiro et al. 1975, Morein et al. 1977, Siwek & Rao 1978, Lotem et al. 1978, Grenier & Guimont 1983, MacEachern & Plewes 1984, Bhole et al. 1985). Most of the patients were elderly and the causes of the tendon ruptures were reported to be degenerative diseases or general metabolic disorders such as gout, obesity, uraemia, hyperparathyroidism or rheumatoid arthritis.

During power lifting, two traumatic cases of knee extensor ruptures have been reported, both at the deepest position of the parallel squat (Grenier & Guimont 1983, Bjerneld et al. 1984) but no biomechanical investigations were made. Zernicke et al. (1977) have performed the only reported biomechanical calculations regarding an in vivo tendon force magnitude at the time of rupture. Their subject was a 29-year-old (82 kg) champion weight-lifter who ruptured his right infrapatellar tendon during a "clean and jerk" lift of 175 kg. The load moment was calculated to 550 Nm and the patellar tendon force to 14,500 N or 17.5 times BW at the time of rupture.

Isokinetic exercise

The isokinetic exercise was first reported in 1967 by Hislop & Perrine and Thistle et al. The isokinetic way of exercise is very efficient in muscle strengthening as the muscle may exert maximum effort throughout the range of motion (*e.g.* Thistle et al. 1967, Pipes & Wilmore 1975, Grimby et al. 1980). In addition, isokinetic exercise is very useful in the study and understanding of biomechanics, kinesiology and physiology because of its standardized motion and its possibility to record directly the joint torque produced. Hence, isokinetic exercise has been widely used for many different purposes for example as a method for analysing force-velocity or power-velocity relationships (*e.g.* Thorstensson et al. 1976, Perrine & Edgerton 1978, Ingemann-Hansen & Halkjaer-Kristensen 1979) or as a training/testing device for healthy athletes (Öberg 1985).

In clinical practice, the isokinetic exerciser has become frequently used for example to assess knee rehabilitation after meniscectomy or ligament injury (Campbell & Glenn 1979, Grimby et al. 1980, Arvidsson et al. 1981, Hamberg & Gillquist 1984, Murray et al. 1984), patello-femoral disorders (Nordesjö & Nordgren 1978, Rauschning et al. 1983, Nordgren et al. 1983, Lysholm et al. 1984), ankle joint function (Nistor et al. 1979, Markhede & Nistor 1979, Fugl-Meyer 1980, Tropp 1985), rehabilitation after femoral shaft fracture (Mira et al. 1980), and spasticity after hemiparesis (Knutsson & Mårtensson 1980). Despite the widespread use of isokinetic exercise, there are no biomechanical reports concerning the forces induced in the knee joint, except for the work by Johnson (1982) who calculated the tibio-femoral shear forces and showed that a dual-pad resistance considerably diminished the stress on the anterior cruciate ligament. However, the compressive force magnitudes were not estimated.

Materials

Subjects

In Study I, fifteen healthy subjects (one woman) volunteered to participate. Mean age was 28 years, weight 76 kg, and height 1.78 kg. During the course of Study I, a new

EMG technique and biomechanical approach were developed. Therefore, only seven subjects were used for moment calculations and five of them for the presentation of EMG recordings.

As subjects in Study II, ten professional milkers (four women) were used with machine milking experience of between 1 and 40 years. The subjects mean age was 31 years, weight 72 kg, and height 1.74 m.

In Studies III and IV, twenty healthy subjects (10 women and 10 men) were radiographed. Their mean age was 23 years (women) and 27 years (men), mean height was 1.67 m (women) and 1.82 m (men), mean weight 59 kg (women) and 75 kg (men). The subjects were without knee disorders or dysfunctions and had not undergone any prior knee surgery.

Four highly-trained power lifters participated in Study V. Their age ranged from 21–34 years, height from 1.64–1.91 m, weight 74–146 kg, and personal record in the parallel squat from 305–430 kg. In a Scandinavian Championship one of the subjects (31 years, 1.91 m, 146 kg) experienced a bilateral simultaneous quadriceps tendon/muscle rupture during a parallel squat while carrying 382.5 kg.

In Study VI, eight healthy men, free from any lower extremity disorder, volunteered as subjects. Their mean age was 27 years, mean weight 72.4 kg, and mean height 1.80 m.

Cadavers

In Studies III and IV, twenty left knees of cadavers (10 women and 10 men) were dissected. Their mean age at death was 76 years (women) and 68 years (men). Their mean height was 1.62 m (women) and 1.74 m (men).

Methods

Calculation of load moments

(Studies I, II, V and VI) In studies I, II and V body position was photographed perpendicularly to the sagittal plane with a motorized camera (Olympus OM-1) at a picture frequency of 4 frames/sec. To make time synchronization possible, a timer with a light-emitting diode display was used. Either photocopies or negative images were projected on a semiautomatic coordinate registration device (Tektronix digitizer, 4953). The digitizer recorded the coordinates of the reference distance points and the positions of the bilateral motion axes of the major joints. The digitizer was connected to a graphics terminal (Tektronix 4012) which was connected to a Nord-100 computer. A sagittal plane mechanical body model based on static mechanics was designed and the load moments about the bilateral knee joint axis and about other major joint axes were calculated (Ekholm et al. 1982). The body segment parameters necessary for the calculations were taken from the anthropometric data given by Dempster (1955), and the moment caused by the burden weight was included in the calculations. The program allowed a change of the burden weight parameter, and this function was used to calculate the contribution of the body segments to the total load moment.

In study VI, a Cybex II isokinetic dynamometer was used connected to an Apple II computer. Controlled acceleration was performed through the computer, and the gravity moments caused by the weights of the leg, foot and resistance arm were corrected for with software (Gransberg & Knutsson 1983).

Electromyographical recordings

(Studies I and II) Muscle activity was recorded by means of EMG from the muscle bellies of the quadriceps and hamstrings muscles. Two flexible disposable Ag-AgCl surface electrodes were attached to the skin over each muscle belly in the direction of the muscle fibers. The distance between the electrode centres was 35–40 mm. The myoelectrical signals were amplified in a Devices AC8 and full wave rectified low pass filtered electromyograms (=linear envelope, Winter 1979) were recorded, using heat-sensitive paper. The bandwidth was 10 Hz to 1.5 kHz and the time integral constant 0.50 sec. For control purposes, amplified unfiltered direct EMG signals were recorded in parallel on a UV-recorder (Honeywell Visicorder, 1508).

For comparison of levels of muscle activity between different muscles and between different individuals, an EMG normalization was performed. The muscle activity was presented as a ratio, TAMP-R, between the EMG activity recorded during the experiments and the activity recorded during an isometric voluntary maximum test contraction. To standardize the isometric test contractions the subject was firmly bound to a specially designed chair and knee angle was held constant at 90 deg for both knee extension and flexion.

Actvities investigated

Lifting (Study I)

A box weighing 12.8 kg placed on the floor in front of the subjects was lifted to a bench at the same level as the subject's umbilicus. Three different lifts were studied; 1) Lift with straight knees (SK), 2) Lift with flexed knees and the box in front of the knees (FKFF), 3) Lift with flexed knees and the box between the feet and thus close to the trunk. All the lifts were performed at a moderate speed, *i.e.* approximately two seconds. To make comparisons possible between the different modes of lifting and between the subjects, time was expressed as a WCR, from 0 to 1. A strain gauge was mounted in each box handle to measure the vertical acceleration of the box and thus obtain an idea of the dynamic force components.

Work postures with hand-held implement (Study II)

Twenty different selected postures were studied in our laboratory at the most loading part of the machine milking procedure, *i.e.* when the teat-cups, weighing 3.3 kg (2.8 kg in the right hand and 0.5 kg in the left hand), were attached to the udder. To simulate pit depth, vertical level differences between milker and cow of 0.17, 0.33, 0.50, 0.70, 0.85, and 1.0 m were arranged. The horizontal distance from the centre of the udder to the milker was 0.50 m. At 1.0 m level difference, horizontal distances of 0.6 and 0.7 m were also studied. Standing postures with the knees straight and, at the lower level differences (0.50 m and less) also with flexed knees, were studied. Nine different postures at zero level-difference were also studied, one of them sitting on a one-legged stool.

Parallel squat (Study V)

In power-lifting, the squat is performed by supporting a weight on the shoulders in a standing position, from which the body is lowered by flexing the knees, so the proximal part of the thighs will become lower than the knees. Three of the four subjects performed parallel squats and the deepest position and the ascending phase were studied. Each subject performed squats in his normal way and speed *i.e.* 1.25 – 1.50 sec for the ascent. Burden weights of between 60 and 260 kg were studied. Using the sagittal plane body model described, the subjects were simulated to the same body size as the subject injured with the bilateral quadriceps rupture.

Isokinetic knee extension (Study VI)
The subjects were seated with back support and the thighs were attached by straps to the table. The arms were held along the trunk and the hands gripped the table sides. The bilateral motion axis about which flexion and extension of the knee occur was aligned with the axis of the resistance arm. The following three knee extensions were conducted in a randomized sequence: 1) velocity of 180 deg/sec, resistance pad distal; 2) velocity of 180 deg/sec, resistance pad proximal; 3) velocity of 30 deg/sec, resistance pad distal.

The distal pad position was just proximal to the ankle joint on the anterior part of the lower at a mean distance of 0.37 m (SD=0.02) from the motion axis. The proximal pad position was 0.20 m from the axis. The knee extension movement started at 90 deg knee angle and finished at straight knee (= 0 deg).

Knee joint model (Studies III and IV)

Dissection
The cadaver's left patella was removed. The thickness of the patellar and quadriceps tendons was measured at their insertions on the patella. The measured distances ap, gt, and bq (Fig. 1) were used to find the tendon midpoints T, P, and Q (Fig. 1) on the radiographs.

Radiography
Lateral x-ray films (30×40 cm) were taken of the loaded knee with the subjects standing at knee flexion angles (α, Fig. 1) ranging from straight (=zero deg) to 120 deg flexed. One anteroposterior and one lateral radiograph picture were taken with the knee straight and the quadriceps activated. The film-focus distance was 1.55 m and the knee centre – film distance was 0.15 m, giving a magnification factor of 1.11 which was corrected for.

The tibio-femoral contact point (C, Fig 1) was defined as the midpoint of the smallest distance between the femoral and tibial condyles. The long axes of the femur and tibia were defined from the radiographs as the lines between the two shaft midpoints located 75 mm and 150 mm from the condyle surfaces.

Forces
Many force vectors act on the bone through one particular tendon. All biomechanical models use approximations, and in this study the forces in the various parts of the tendon were presumed to act as one total single force vector from tendon midpoint at one place to tendon midpoint at the other. The same approach in biomechanical modelling has been used earlier (Jensen & Davy 1975, Dostal & Andrews 1981, Németh & Ohlsén 1985). This assumption is fair for straight tendons, but at knee angles of 90 deg or more where the quadriceps and patellar tendon are no longer straight, two extra points (E and K) were defined (Fig. 1). Point E was locted on the radiographs as the midpoint of the quadriceps tendon, 15 mm proximal to the patellar base. Point K was located at a distance corresponding to half patellar tendon thickness in front of the anterior border of the tibial plateau, and the direction of the patellar tendon force (Fp)

was defined as the straight line between points K and P.

The compressive force in the patello-femoral joint (Fcp) was projected through the centre of the patello-femoral contact point (M), perpendicularly to the joint surface (Fig. 1). The force in the quadriceps tendon (Fq) was assumed to run parallel to the long axis of femur at knee angles of 30 deg and 60 deg. Fq passes through the midpoint of the quadriceps tendon insertion at the patellar base (point Q). To find point Q on the x-rays, the proximal dorsal bony corner of the patellar base (point B) was identified. The distance between these two points, bq-distance, was estimated in the morphological study and its mean value was used on the x-rays. The corresponding approach was used to locate the tendon midpoints at patellar apex (P) and at tuberositas tibiae (T). At the straight knee the direction of Fq was determined as the line between the Fp – Fcp intersection point (X) and the midpoint of the quadriceps tendon (point Q) at the patellar base.

At knee angles 90 deg and 120 deg the quadriceps tendon was curved in the femoral intercondylar groove. The force magnitude in the proximal part of the quadriceps muscle (Fqm) was considered to be the same as in the distal part (Fq) inserting on the patellar base. The angle between these two forces was defined the λ angle. It gave rise to a force between the distal part of the quadriceps tendon and femoral intercondylar groove (Fcq), (Fig. 1). At knee angles above 60 deg, the direction of Fq was estimated as the line between points E and Q (Fig. 1).

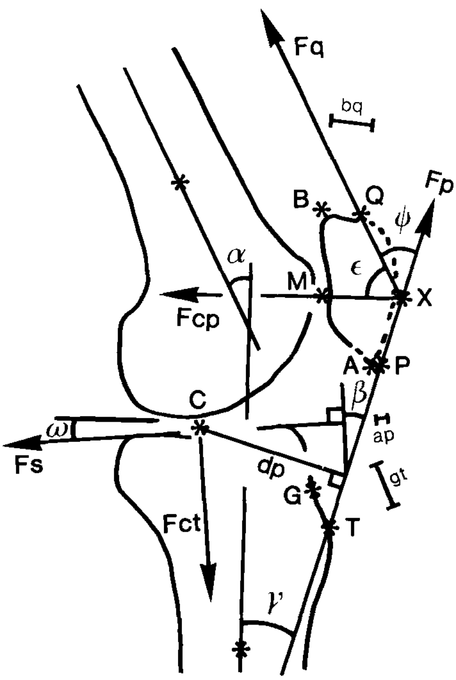

Fig. 1. Forces, angles, distances and bony landmarks at 30 deg knee angle (α). The patellar tendon moment arm (dp) is the perpendicular distance between patellar tendon force (Fp) and centre of the tibio-femoral contact (C). Fp acts through proximal patellar tendon midpoint (P) with a distance "ap" from patellar apex (A) and through distal midpoint (T) with distance "gt" from posterior tendon border (G). Tibial plateau slopes backwards (ω, women 7.2 deg and men 9.2 deg) in relaticn to tibial long axis. The tibio-femoral forces Fct and Fs act through C perpendicularly to and along the joint surface, respectively. Quadriceps tendon force (Fq) parallels femoral long axis and goes through tendon midpoint (Q) at the insertion on patellar base. Patello-femoral compressive force (Fcp) projects through Fp-Fq intersection point (X) and centre of patello-femoral contact point (M), perpendicularly to the joint surface.

Mechanical analysis
The tibio-femoral contact point (C) was chosen as the origin for the calculation of moments. For the tibia in a free body diagram and in a static situation (Fig. 2a), the moment (M) about C can be determined:

$$M = (de \times Fet) + (df \times Fef) + (dm \times mg) \tag{1}$$

where Fet, Fef and mg are the "external" forces acting on the tibia and de, df, and dm their moment arms. The force in the patellar tendon (Fp) is given by:

$$Fp = M / dp \tag{2}$$

where dp is the moment arm of Fp. From Fig. 2a, we get the vector equation:

$$\mathbf{Fp} + \mathbf{Fct} + \mathbf{Fs} + \mathbf{Fet} + \mathbf{Fef} + \mathbf{mg} = \mathbf{0} \tag{3}$$

Projecting (3) in the normal (or Fct) direction of the tibial plateau;

$$Fct = Fp \times \cos\beta + Fet \times \sin\delta 1 + Fef \times \sin\delta 2 - mg \times \sin\delta 3 \tag{4}$$

in tangential (or Fs) direction;

$$Fs = Fp \times \sin\beta - Fet \times \cos\delta 1 - Fef \times \cos\delta 2 - mg \times \cos\delta 3 \tag{5}$$

where β, δ1, δ2 and δ3 are angles in relation to tibial plateau. In the example above the external forces act on the anterior border of tibia and on the sole of the foot, but it is obvious that these forces may act anywhere on the "free body".

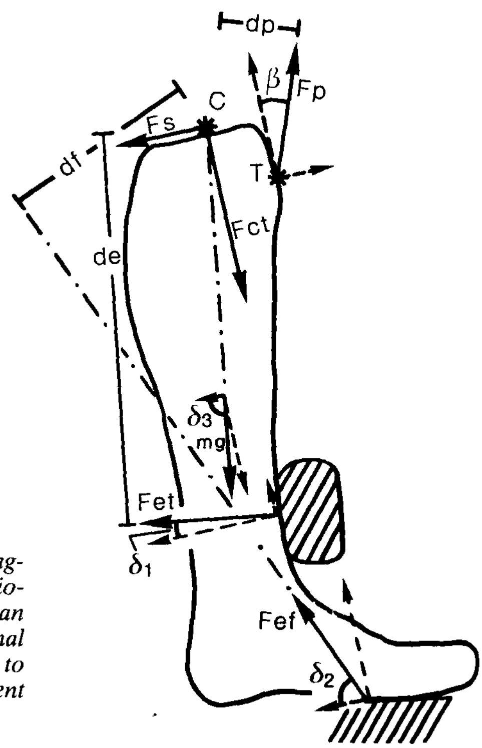

Fig. 2a. Free body diagram of lower leg. The magnitudes of the patellar tendon force (Fp), tibio-femoral shear (Fs) and compressive force (Fct) can be calculated (see equations above) if the external forces (Fet, Fef and mg), their angles in relation to the tibial plateau (δ1, δ2 and δ3) and their moment arms (de and df) have been determined.

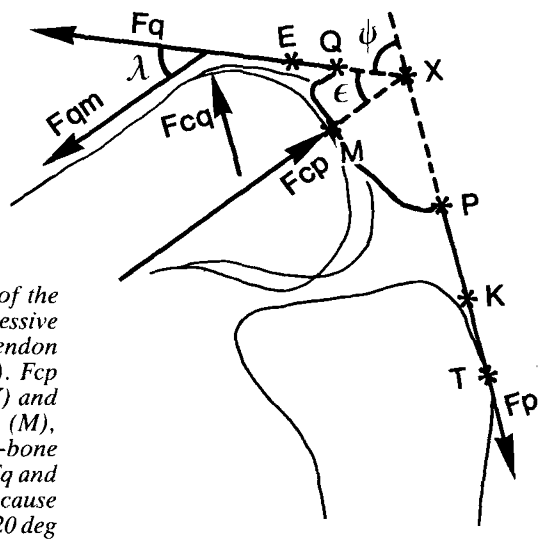

Fig. 2b. Free body diagram of patella and of the quadriceps tendon. Patello-femoral compressive force (Fcp) equals the vector sum of patellar tendon force (Fp) and quadriceps tendon force (Fq). Fcp projects through Fp-Fq intersection point (X) and centre of patello-femoral joint contact point (M), perpendicularly to the joint surface. Tendon-bone compression (Fcq) equals the vector sum of Fq and Fqm. Tendon midpoints E and K are used because the knee extensor tendon is curved at 90 and 120 deg knee flexion angles.

Three forces act on the patella. If we consider the patella in a free body diagram (Fig. 2b) we get the vector equation:

$$\mathbf{Fp} + \mathbf{Fq} + \mathbf{Fcp} = \mathbf{0} \qquad (6)$$

From basic trigonometry we know that:

$$\mathrm{Fcp} = \mathrm{Fp}\,(\sin \psi)\,/\,(\sin \varepsilon) \qquad (7)$$

Projecting the vectors Fcp and Fp on Fq gives the magnitude of Fq:

$$\mathrm{Fq} = \mathrm{Fcp}\,(\cos \varepsilon) + \mathrm{Fp}\,(\cos \psi) \qquad (8)$$

Considering the distal part of the quadriceps tendon in a free body diagram we get the vector equation:

$$\mathbf{Fq} + \mathbf{Fqm} + \mathbf{Fcq} = \mathbf{0} \qquad (9)$$

As forces Fq and Fqm are assumed to have the same magnitude we get from equation (9):

$$\mathrm{Fcq} = 2\,\mathrm{Fq}\,(\sin \lambda/2) \qquad (10)$$

Statistics

Two- and three-way analysis of variance (ANOVA) were used in Studies I-IV. Multiple comparisons using studentized range test ($p = 0.01$) were used if ANOVA showed significant difference at 0.01 level. In studies III, IV and VI, Student's unpaired or paired t-tests were used. The significance levels chosen were consistently $p = 0.01$.

Means with 95 % confidence intervals were calculated in Studies I-IV and means with standard deviations in Study VI. For the EMG-recordings in Study II, the medians with their 95 % confidence intervals were calculated.

Summary of Results

Lifting (Study I)

Lifting the box with straight knees (SK) gave throughout the lift an extending knee load moment which was highest at the beginning and at the end of the lift (Fig. 3). In the middle of the lift (WCR=0.6) the load was significantly lower. Lifting with flexed knees (FKFF and FKC) induced a flexing knee moment of about 50 Nm at the beginning of the lift, and during the course of the lift the moment changed to an extending direction. There was no significant difference in load moment between the two lifts with flexed knees. During the final phase (WCR 0.8 and 1.0) of all lifts, there was no load difference between the lifts, and the extending knee moment was about 55 Nm. During the SK-lift, the knee extensor muscle showed low activity throughout. The hamstrings, which are knee flexors and also hip extensors, were activated to a medium level during the first third of the lift. During the flexed knee lifts, there was medium to high activity level in the knee extensors. Although to a low level, the hamstrings were activated, indicating that a certain co-contraction was present.

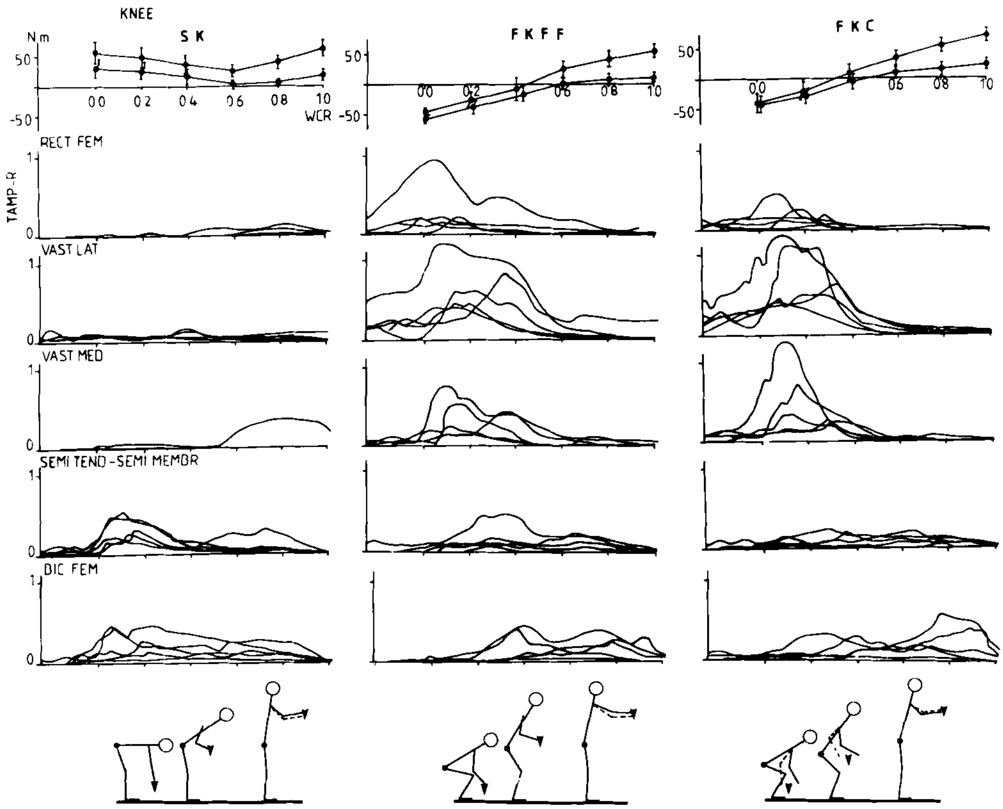

*Fig. 3. **Diagrams above:** load moment (y-axis) about bilateral knee axis during the course (x-axis, WCR) of the three different lifts SK, FKFF and FKC. Positive values show extending load moment, negative flexing load moment. Upper curve: total load moment. Lower curve: moment due to body segments only (burden subtracted). Means and 95 % confidence intervals, n = 7. **Diagrams below:** level of EMG-activity in TAMP-R in five knee muscles, n = 5.*

Work postures with hand-held implement (Study II)

In the straight knee postures, a level difference of 1.0 m between milker and cow showed the lowest extending moment (Fig. 4). At small level differences (0.0 m – 0.50 m) the knee moment was significantly higher, and some subjects utilized a high percentage of the muscle capacity. Increased horizontal distance between milker and cow induced higher knee extending moments. In flexed knee postures the knee load was of flexing direction, and the vastus lateralis muscle was activated between medium and high level and more than the rectus femoris muscle. In both straight- and flexed-knee postures, the biceps femoris muscle was activated between low and medium levels. A knee flexion angle of 65 deg gave a knee moment close to zero, regardless of body weight.

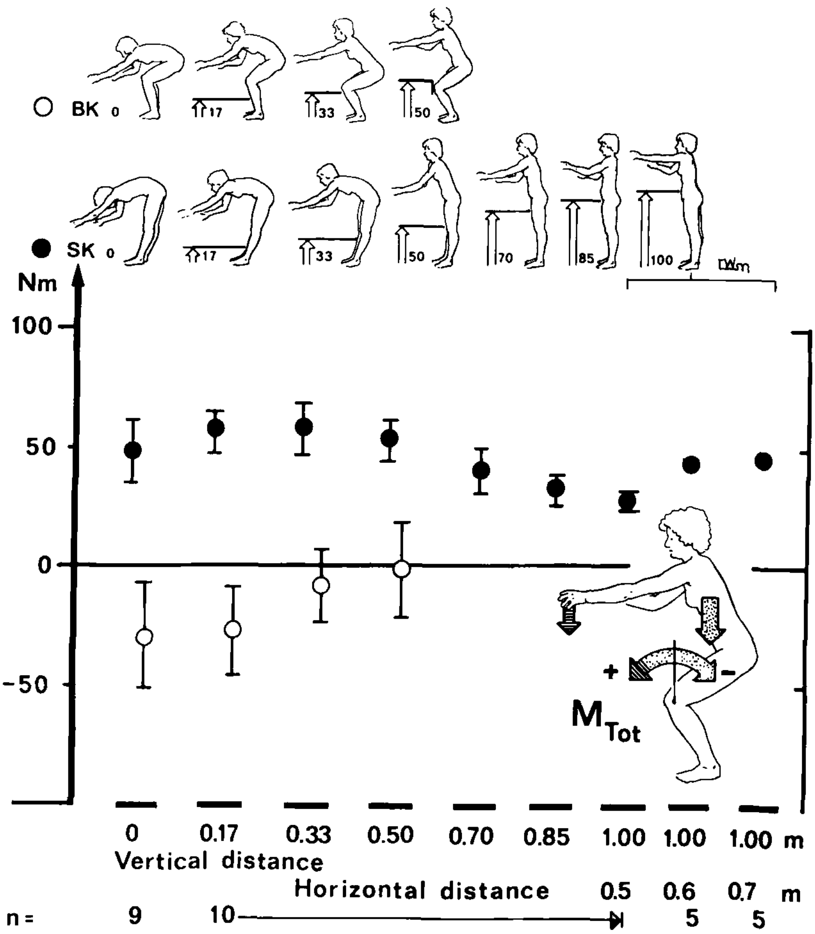

Fig. 4. Knee load moment in different standing working postures (x-axis) illustrated above. Filled circles = straight knees. Unfilled = flexed knees. Positive values indicate extending knee load moment, negative values flexing moment. The figures on the x-axis indicate different level differences between operator and cow. At 1,0 m three different horizontal distances between cow and operator were studied. 95% confidence intervals of the means. n = 9 for 0.0 m. n = 5 for 1,0 m with horizontal distance 0,6 and 0,7 m. n = 10 for the remaining postures. Reprinted with kind permission from Ergonomics (Study II); copyright 1985 by Taylor & Francis Ltd.

Knee joint model (Studies III and IV)

Tibio-femoral contact point

The tibio-femoral contact point (C) moved anteriorly on the tibial plateau when the knee was extended (Fig. 5). Over the last 30 deg of knee extension, point C was much displaced. The displacement of C during extension from 120 deg to straight was about 20 mm, or 40 % of the tibial plateau sagittal length.

Comments. The displacement pattern of point C indicates that a considerable rolling motion occurs in the tibio-femoral joint during the last deg of knee extension. The results are mainly in agreement with earlier reports (Zuppinger 1904, Lindahl & Movin 1967, Walker & Hajek 1972). However, this study showed a tibio-femoral rolling motion also at more flexion angles greater than 30 deg. A point C displacement of as much as 20 mm has not earlier been reported.

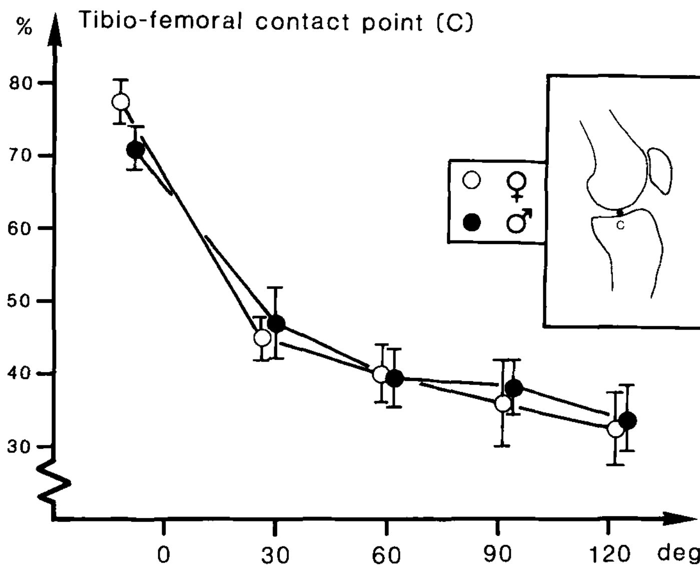

Fig. 5. Location of tibio-femoral contact point (C, y-axis) at various knee flexion angles (α, x-axis). 95 % confidence intervals of the means are indicated. Y-axis shows the displacement in % of the sagittal tibial plateau length. 100 % equals the anterior border of the plateau.

Patellar tendon angle

The magnitude of the patellar tendon angle γ varies with knee angle as shown in Fig. 6a. The horizontal axis shows the radiographically determined knee angle (α) which for the straight knee was more negative for women (mean -11.7 deg) than for men (-7.8 deg). The angle β is of greater biomechanical interest than γ, as β represents the patellar tendon direction in relation to the tibial plateau. β equals the sum of γ and ω. The magnitude of β increases in the same way as γ during knee extension. These two angles change from a negative magnitude to a positive at a knee angle of 75 and 100 deg, respectively.

Comments. The magnitude of γ at various knee angles has been reported by Smidt (1973). Smidt's pattern of change is similar to that in the present study but consistently somewhat higher. The magnitude of β at various knee angles is of importance when

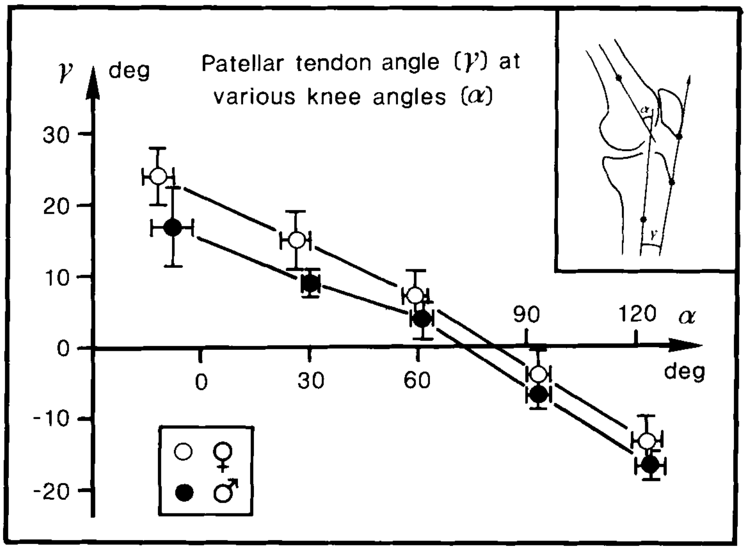

Fig. 6a. Magnitude of γ (y-axis) at various knee flexion angles (α, x-axis). 95 % confidence intervals of the means are drawn for both γ and α. Zero deg α equals straight knee. The difference in γ is statistically significant between men and women.

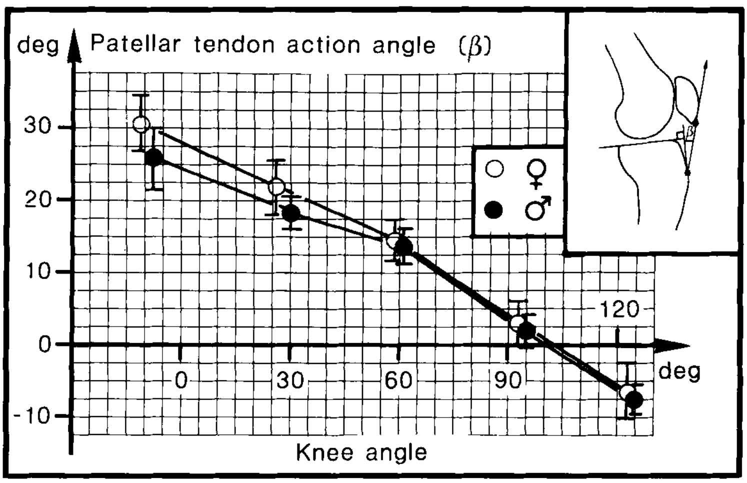

Fig. 6b. Magnitude of β (y-axis) at various knee flexion angles (α, x-axis). 95 % confidence intervals of the means are indicated. Note the intersection of the zero-line at about 100 deg knee angle.

constructing a local biomechanical model of the knee but has not earlier been determined. When the knee is straight the patellar tendon pulls tibia forwards in relation to femur due to β being positive and when the knee is flexed to more than 100 deg, the patellar tendon pulls tibia backwards.

Patellar tendon moment arm
The maximum length of the patellar tendon moment arm was found at 30 to 60 deg knee flexion angle, and the shortest at the straight knee (Fig. 7). The same pattern was seen for both sexes but women had significantly shorter moment arms than men throughout the whole range of motion.

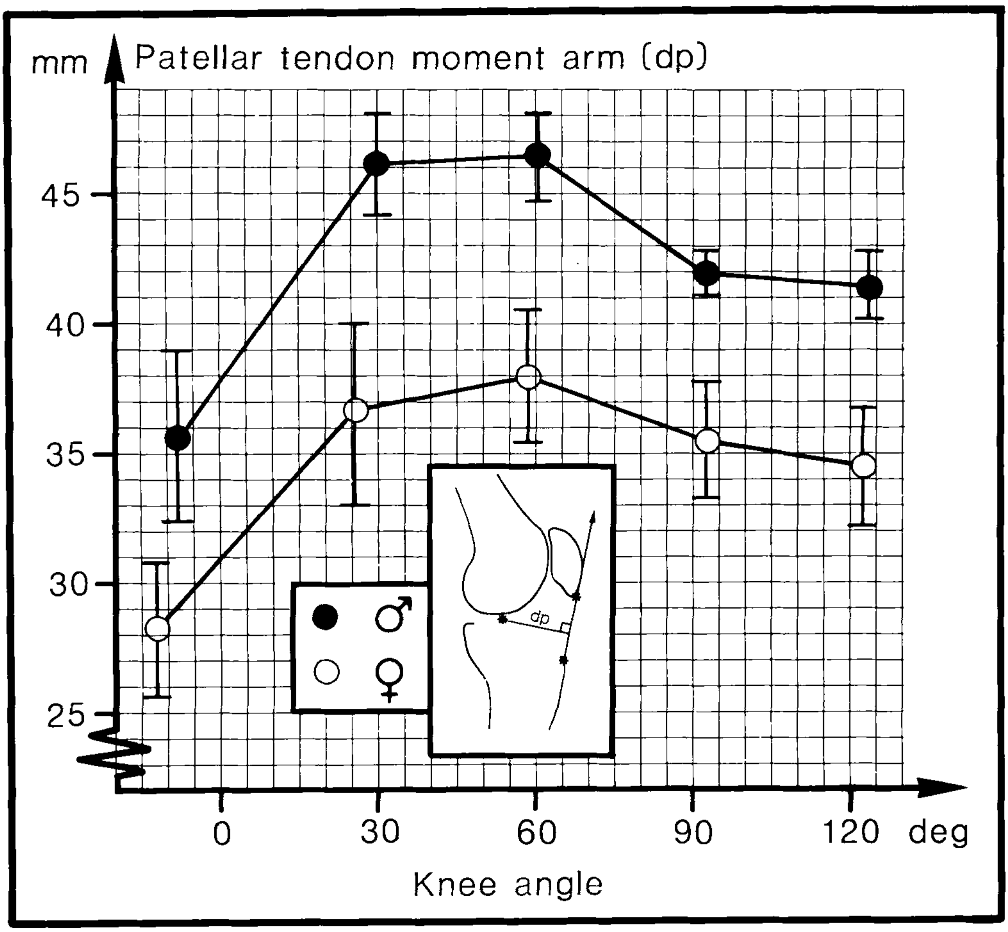

Fig. 7. Patellar tendon moment arm (dp, y-axis) for men and women at various knee flexion angles (x-axis). 95 % confidence intervals of the means are indicated.

Joint forces

Fig. 8 shows an isometric knee extension against a resistance, Fet, applied at a distance, de, perpendicularly to the anterior side of the distal leg, with the subject lying on one side. As women had shorter moment arms than men (Fig. 7), the calculated forces for women were consistently about 20 % higher. The patellar tendon force (Fp) and tibio-femoral compressive force (Fct) were of the same magnitude and did not change dramatically over the various knee angles. When the knee was extended by a moment of 40 Nm, it was found that Fp and Fct were about 1100 N at knee angles of 30–120 deg (Fig. 8 a+b). For the straight knee these forces were higher, about 1230 N and 1400 N respectively. The tibio-femoral shear force (Fs) changed sign between 50 and 90 deg flexion angle, depending on the magnitudes of Fet and de (Fig. 8c). Fs was around −200 N at 120 deg knee angle and as much as 600 N anteriorly shearing when the knee was straight.

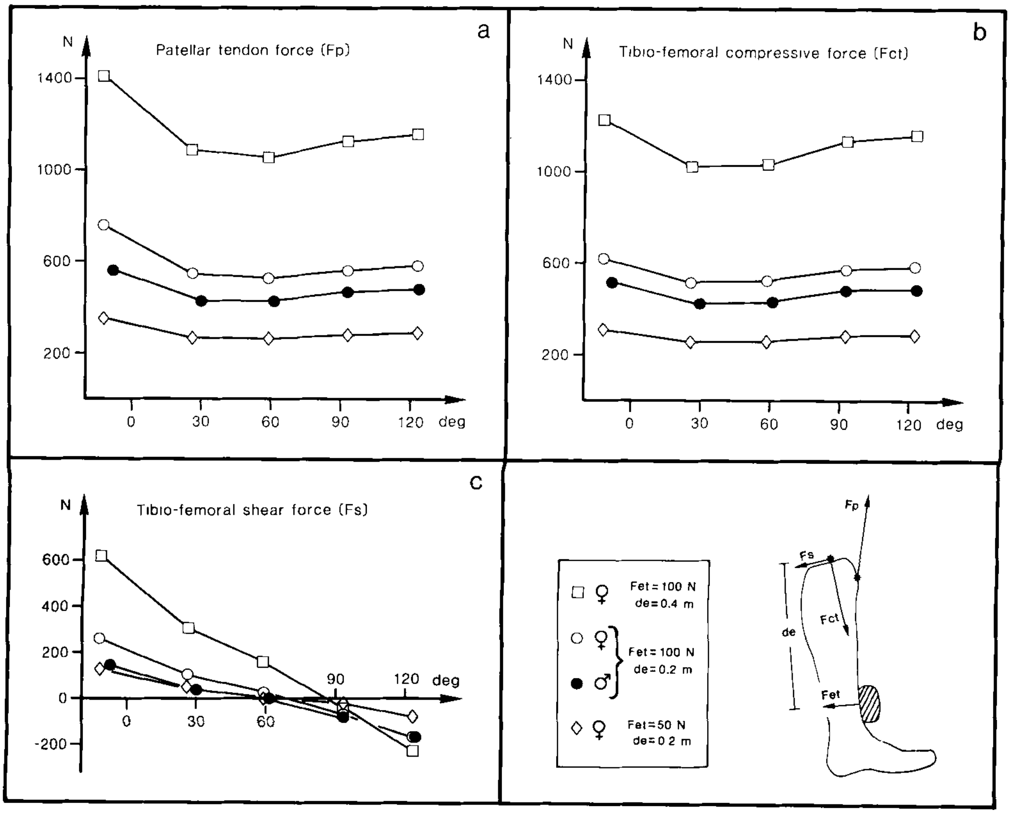

Fig. 8. a) Patellar tendon force (Fp), b) tibio-femoral compressive force (Fct) and, c) tibio-femoral shear force (Fs) for women (unfilled symbols) and men (filled circles) during isometric knee extension at various knee angles (x-axis). The external force (Fet) and its varying moment arm (de) are indicated. Note in a) and b) that the same magnitudes of Fet and de give higher forces for women than for men. Note in c) that the intersection of the Fs zero line is between 50 and 90 deg knee angle. Positive values in c) indicate that tibia tends to shear anteriorly in relation to femur, negative values – posteriorly shearing.

The magnitudes of the patello-femoral compressive force (Fcp) and tibio-femoral shear force (Fs) were determined during knee extension against gravity with and without a 31 N boot (Fig. 9). Fcp reached its maximum at 50 deg knee angle, and at straighter knee angles with higher extending muscle moment this force declined. The anteriorly directed Fs reached its maximum during the last few degrees of knee extension.

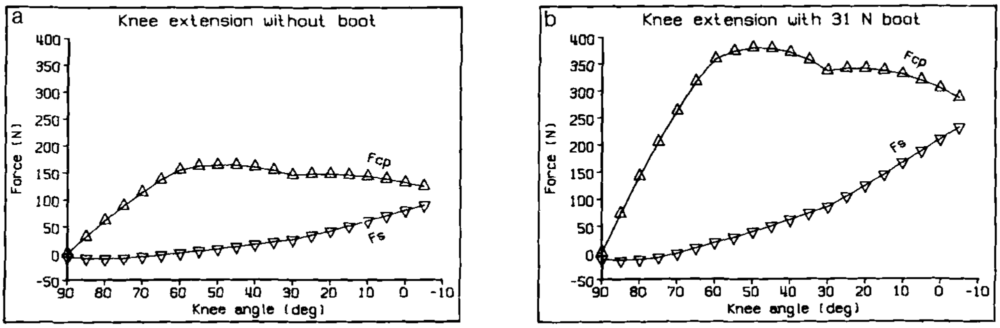

Fig. 9. Calculated patello-femoral joint compressive force (Fcp) and tibio-femoral shear force (Fs) during knee extension from 90 to zero deg (x-axis). a) Without boot, b) With a 31-N-boot.

The patellar forces may be calculated using equations (2), (7), (8), and (10). In Fig. 10c it is seen that the magnitude of the patello-femoral compressive force (Fcp) reached its maximum at 90 deg knee angle and decreased slightly towards 120 deg. The patellar tendon force (Fp) was lower than the quadriceps tendon force (Fq) at knee flexion angles of 60 deg to 120 deg and the minimum Fp/Fq ratio was around 0.70. A ratio of about 0.70 was also seen when comparing the patellar tendon thickness with that of the quadriceps tendon. The magnitude of the compressive force between the quadriceps tendon and the femoral intercondylar groove (Fcq) increased considerably at knee flexion angles more than 60 deg.

Comments. Note that the patellar forces can be quantified by using the force prediction diagram (Fig. 10) if three variables are known; 1) *sex*, 2) *knee moment* and, 3) *knee flexion angle*.

Quadriceps rupture (Study V)

The forces in the knee joint at the situation of complete bilateral quadriceps muscle rupture were calculated. The force in the quadriceps tendon (Fq) was greater than the other forces calculated, and its magnitude was between 10,900 N and 18,300 N. The flexing knee load moment ranged between 335 Nm and 550 Nm while the maximum flexing moments about the hip (410–670 Nm) and the lumbo-sacral joint (575–1100 Nm) were even greater.

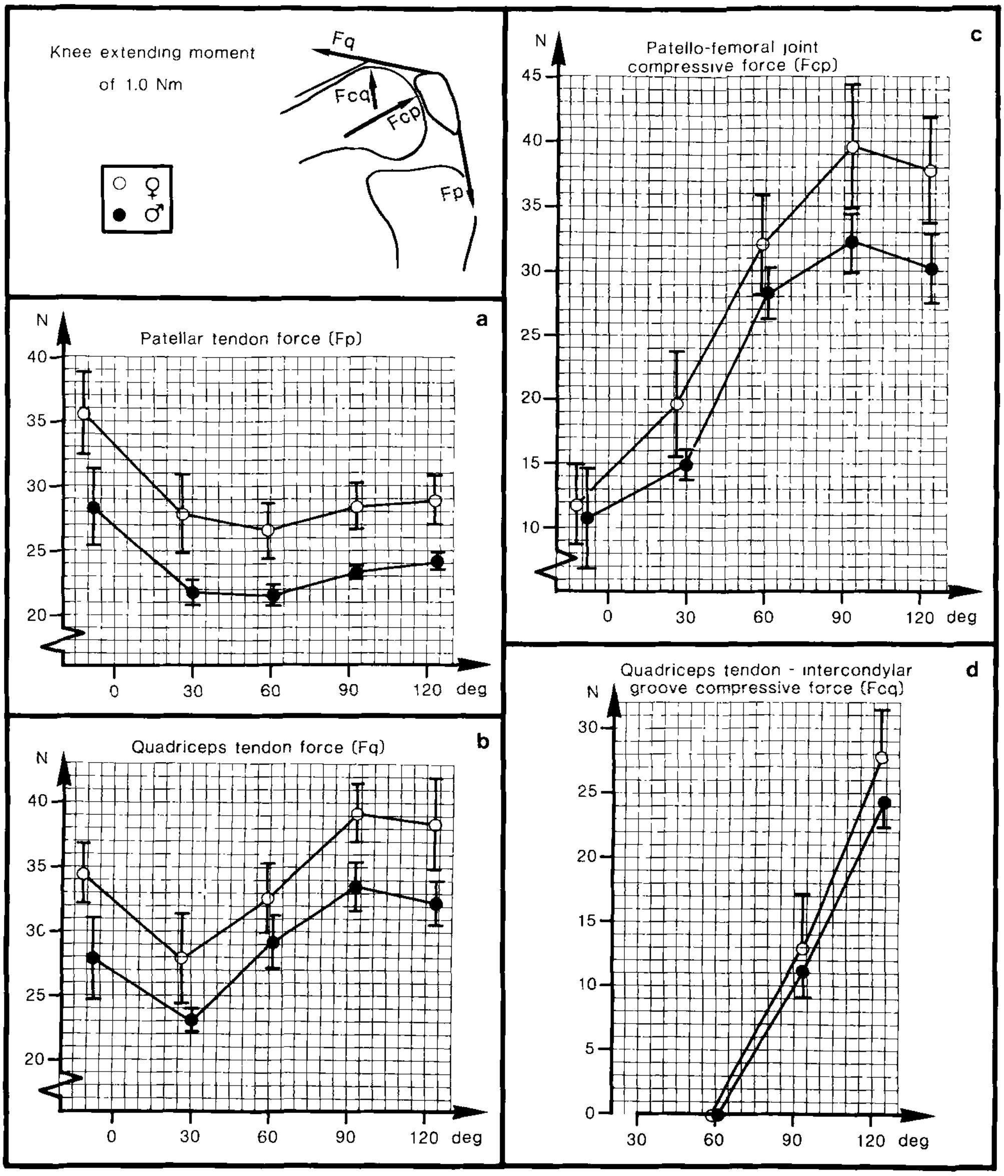

Fig. 10. Prediction diagram for quantifying forces in a) patellar tendon, b) quadriceps tendon, c) patello-femoral joint and d) between quadriceps tendon and intercondylar groove at various knee flexion angles (x-axis) for women (unfilled circles) and men (filled circles). 95 % confidence intervals of the means are indicated. Let us look at one example of how Fcp can be determined. Consider a load moment of 210 Nm and a knee angle of 50 deg (as during jogging, Winter 1983). Enter Fig. 10c at the horizontal axis at 50 deg, continue vertically to the lower curve (male subjects), then horizontally to the vertical axis and read "24 N", which is multiplied by the moment value "210". This gives the force magnitude 5040 N, which is above 7 times BW for a 72 kg man. The other patellar forces can be calculated in a corresponding way. Reprinted with kind permission from the Scandinavian Journal of Rehabilitation Medicine (Study IV).

Isokinetic knee extension (Study VI)

At 180 deg/sec, the maximum moments produced by the quadriceps muscle reached mean peak magnitudes of 181 Nm at 65 deg knee flexion angle with the resistance pad placed distally. With the pad placed proximally, the peak moment occurred at the same knee flexion angle but was significantly lower (163 Nm). At 30 deg/sec, the mean magnitude of the peak moment was significantly higher (284 Nm) than at the faster speed, and the peak occurred somewhat earlier in the knee extension movement, at 70 deg knee flexion angle.

The tibio-femoral shear force (Fs) changed direction from being negative (tibia tends to move posteriorly in relation to femur) at the starting position, to a positive value when the knee was extended. Having the resistance pad proximal gave significantly lower anteriorly-directed shear forces but, on the other hand, higher posteriorly-directed shear forces. Knee extension of 30 deg/sec induced the highest anteriorly-directed shear force, with maximum mean magnitudes close to 1 BW (= 710 N), seen between 60 and 40 deg knee flexion angle.

Comments. Johnson (1982) noted also that the maximum moments induced by the knee extensor muscles become lower when the resistance pad is placed proximally on the leg. The reason for this difference is obscure but may possibly be an unfamiliar or uncomfortable feeling experienced when the pad is positioned proximally. The higher resistance forces may induce pain or other inhibiting influences from the cutaneous, periosteal and/or ligamentous mechanoreceptor afferents, which reflexly inhibit the activation of the motorneurons (Ekholm et al. 1960). Another explanation would be that the higher resistance force influences the tibial position in relation to femur (the proximal end of the tibia is kept more backwards), making the patellar tendon moment arm shorter. If this is true, the same maximum forces exerted by the quadriceps muscle and patellar tendon would produce *lower* knee extending moments due to an impaired mechanical geometry.

Discussion

Assumptions of the knee model

The two-dimensional local biomechanical knee model was made possible by using dissection and radiography in combination, and applying the anatomical findings to biomechanical principles. However, for a complete picture of the forces transmitted in the knee, a three-dimensional force analysis would be needed. Furthermore, the present model did not take into account any antagonistic muscle activity neither from hamstrings nor gastrocnemius muscles. In Studies I and II it was seen that the hamstrings were activated together with quadriceps action. The same has been shown during other knee extending activities when the foot is "fixed", either to the ground as in standing positions or in a closed chain as in bicycling (*e.g.* Carlsöö & Molbech 1966, Kelley et al. 1978, Schüldt et al. 1983, Németh et al. 1984, Gregor et al. 1985, Ericson et al. 1985a). However, during knee extension with the foot not fixed – as during isokinetic knee extension – antagonistic muscle activity seems to be low (Osternig et al. 1984).

If the antagonistic muscle forces had been considered, the calculated joint force magnitudes would have been even higher. *However, there are no biomechanical models that consider antagonistic muscles, since the major problem is to quantify their tendon forces.* With electromyography it is possible to study and quantify the activity of such antagonistic muscles but difficult to quantify the tendon force magnitudes because the relation between muscle activity and tendon force is not always linear (Jonsson 1978, Woods & Bigland-Ritchie 1983).

Using a permutation technique, Nissan (1980) showed the sensitivity of several variables in knee joint biomechanics. He concluded that accurate anteroposterior localisation of the contact point between tibia and femur is most important in the analysis of the knee joint. If the tibio-femoral contact point were moved backward 10 mm in our model, the calculated force in the patellar tendon would decrease by a maximum of approximately 22 %. If the contact point were moved 10 mm anteriorly this force would increase up to 40 %. Since tibio-femoral contact changes with knee angle, especially towards the end of knee extension, *it is important to identify the location of the contact point accurately in order to get proper data of the knee extensor moment arm and, thus, of the forces calculated.*

In the present study, the moment point chosen was the centre point of contact (C) between the condyles. It is a mathematical advantage to choose the contact point as the moment point because the compressive force acts through it and thus gives rise to a zero moment. The shear force is balanced by structures with short or zero moment arms to the contact point (such as cruciates, menisci, collateral ligaments and joint capsule) and therefore it causes a small or zero moment about this point. *In models that use the centre of the axis as the moment point (e.g. Smidt 1973), the moments caused by the joint compressive and shear forces have been omitted, which may give incorrect values of the force magnitudes calculated.*

With regard to the assumptions discussed above, the knee biomechanical model is considered as a useful instrument for quantifying knee joint forces. Comparing results from the biomechanical model with experimentally measured forces in the anterior cruciate ligament (Paulos et al. 1981, Arms et al. 1984), and in the quadriceps tendon (Grood et al. 1984), gives similar patterns and magnitudes of the force curves. In addition, experimentally measured relations between patellar (Fp) and quadriceps tendon forces (Fq) (Haxton 1945, Bishop & Denham 1977, Ellis et al 1980, Huberti et al. 1984) correlate well with the relation found from the present model. Furthermore, experimentally measured patello-femoral compressive forces (Ahmed et al. 1983, Huberti & Hayes 1984) have magnitudes and change patterns similar to the results from the present model. *These facts indicate that the model should be valid.*

Compressive force per area

When considering the magnitude of the tibio-femoral compressive force (Fct), it should not be forgotten that this force acts through an area which varies with knee angle (Kettelkamp & Jacobs 1972, Maquet et al. 1975, Seedhom & Hargreaves 1979, Fukubayashi & Kurosawa 1980, Ahmed & Burke 1983). The tibio-femoral contact area is larger at straighter knee angles, indicating that tibio-femoral pressure is lower for straighter knees for the same Fct magnitude. In Fig. 8 it is seen that Fct reaches its maximum magnitude when the knee is straight but the tibio-femoral pressure is not highest in this position. Using the contact area magnitude found by Ahmed & Burke (1983), the area is about 50 % higher at zero deg than at 90 deg knee flexion angle. Thus, the tibio-femoral pressure calculated with the present biomechanical knee model (Fig. 10) should for the straight knee be about 25 % *lower* than for the 90-deg-flexed, although the compressive force is somewhat higher.

The same discussion applies to the patello-femoral joint since, here also, the contact area varies with knee joint angle. The largest area is reported at 60–90 deg knee flexion angle and the smallest when the knee is straight (Matthews et al. 1977, Ficat & Hungerford 1977, Huberti & Hayes 1984). From this follows that *the patello-femoral pressure, also when the knee is almost straight, may be considerable although the compressive force seems to be comparatively low.*

Knee joint loads during various activities

In Table I the maximum knee load moments and patello-femoral compressive forces (Fcp) during various activities are presented. The knee moments were taken both from the literature and from Studies I, II, V and VI. Fcp was calculated using the "force prediction diagram" presented in Fig. 10 since the three varibles sex, knee moment and knee angle were known. During level walking the flexing knee load moment ranges between 35 and 61 Nm and Fcp is not above 850 N. During jogging, descending stairs, kicking or rising from a chair, the knee moment is considerably higher and the knee flexion angle more pronounced which induce much higher Fcp magnitudes.

TABLE I. Maximum patello-femoral compressive force (Fcp) and knee extensor moments (Mk) produced by quadriceps muscle during various activities. The number of female subjects in the different experiments is given in brackets. Fcp was calculated using the force diagram presented in Fig. 10 and, as there is a difference in this diagram between men and woman, the sexes were considered.

Author(s)	Activity	Number of subjects	Mean weight (kg)	Knee angle (deg)	Mk (Nm)	Fcp (N)
Bresler & Frankel 1950	Level walking	4	71	20	60	840
Radcliffe 1962	Level walking	4	–	20	61	850
Morrison 1968	Level walking	1	–	–	35*	490
Boccardi et al. 1981	Level walking	1	65	–	60*	840
Winter 1980	Level walking	1(1)	–	–	45*	830
Winter 1983	Jogging	11	72	50	210	5000
Andriacchi et al. 1980	Ascending stairs	10	71	65	54	1500
Andriacchi et al. 1980	Descending stairs	10	71	60	147	4000
Lindahl et al. 1969	Isometric max	12	–	60	225	6100
Smidt 1973	Isometric max	26	82	60	120	3400
Wahrenberg et al. 1978	Kicking	6	76	100	180	5800
Kelley et al. 1978	Rising from chair	6(3)	–	90	110	3800
Schüldt et al. 1983	Rising from squat	9(5)	63	105	70	2500
Ericson et al. 1985b	Bicycling	6	71	29	79	880
Study I	Lifting	7	77	90	50	1600
Study II	Machine milking	10(4)	72	108	31	1100
Study II	Isometric max	10(4)	72	90	198	6900
Study V	Parallel squat	3	146**	120	455	14900
Study VI	Isokinetic knee ext	8	72	70	284	8300

* Knee flexion angle of 20 deg was used

** The body weight was simulated

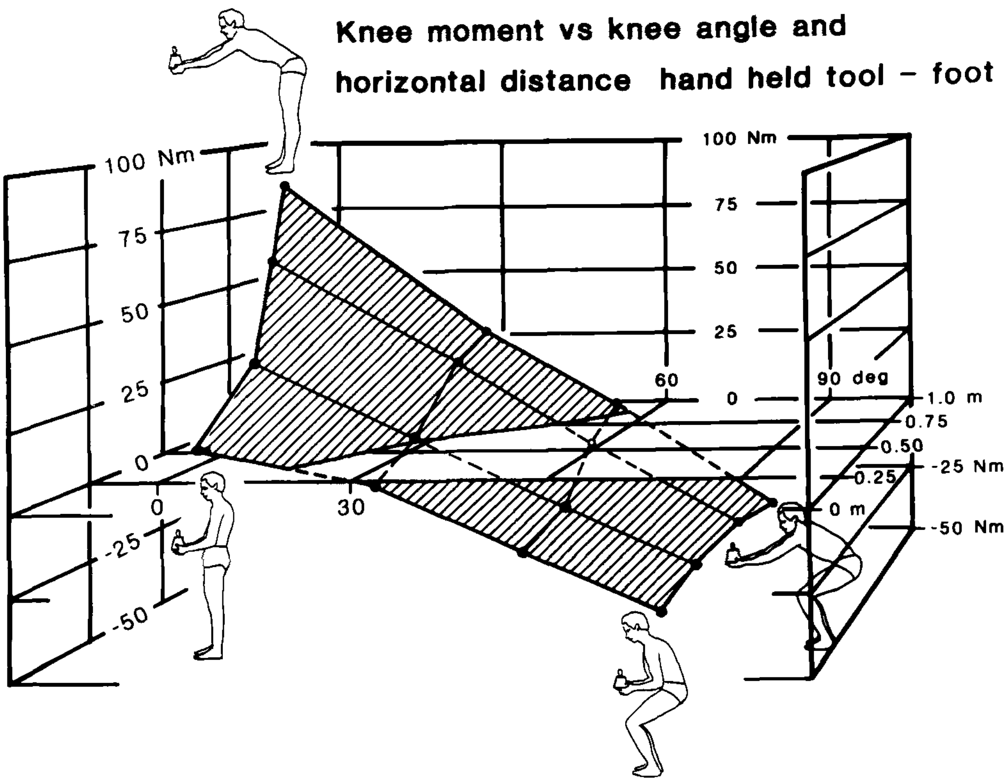

Fig. 11. Knee load moment (y-axis, Nm) for one subject (72 kg) at various knee angles (x-axis, deg) and at various horizontal distances between hand held equipment (3.3 kg) and ankle joint (z-axis, m). The illustrations show the body positions at the four "corners". The zero knee load moment is the intersection of the diagram at the x-z plane.

Work postures

The use of the sagittal plane body model presented makes it possible to quantify knee joint loads in terms of load moments and also to compare different work postures or working tasks with each other. It is important to remember that to find the "best" work posture is to minimize the loading effect in a whole-body perspective and therefore all major joints must be considered. Studies I and II have focused only on the knee joint and the optimum working situation – with respect only to the knee joint – is found when the knee moment and muscle activity are low, which give low compressive and tensile forces in and around the joint. The lowest knee load calculated during machine milking (for the standing low level difference postures) occured when the knees were flexed to about 65 degrees. In this position the knee load moment was around zero.

The sagittal knee load moment – in static or slow motioned standing activities such as those studied in the present work – is dependent on four parameters; 1) *body weight*, 2) *knee angle*, 3) *horizontal distance* to the working area, 4) *burden weight*. Fig. 11 shows the loading effects of changing two of these parameters, namely knee angle and horizontal distance. The most extending knee load moment occurs when the knees are straight and the horizontal distance to the working area is pronounced (top left in Fig. 11). If the hands with the burden move closer to the body (and the knees are still straight) the extending knee moment decreases (lower left). If the knees start bending from this position the knee load changes to a flexing moment, which is maximum when the burden is close to the body and the knees most flexed. The flexing knee load moment decreases if the burden moves away from the trunk (to the very right) and changes to an extending moment when knee straightening occurs back to the top left position.

Thus, it is possible to determine the sagittal knee load moment when the parameters described here are known. The same principles can be used for determining knee moments in other similar standing work situations. In most kinds of lifting work and in many kinds of repair work this model can be generally used in order to find the least load on the knee joints.

Clinical implications

The pattern of change in patellar tendon moment arm (Fig. 7) was quite similar to that in earlier reports (Haxton 1945, Lindahl & Movin 1967, Kaufer 1971, Bandi 1972, Smidt 1973). The main difference between the present results and others is seen at straighter knee angles, where the moment arm was suggested to be about 10 mm shorter than what has been reported earlier. This fact, together with the physiological muscular length-tension relationship (Wilkie 1956, Lunnen et al. 1981), may be one explanation why knee extending strength decreases during the last 30–40 deg of knee extension (Williams & Stutzman 1959, Hallén & Lindahl 1967, Lieb & Perry 1968 and 1971, Lindahl & Movin 1968, Lindahl et al. 1969, Moritz et al. 1973, Smidt 1973). *It may also explain the difficulty, often seen in clinical practice, in achieving the last few degrees of knee extension.*

Due to differences between men and women in length of the patellar tendon moment arm (Fig. 7), the female hyaline cartilage will be exposed to higher forces if the extending muscular moment is of the same magnitude. In addition, the contact area over which the compressive force acts, is smaller in women than in men due to smaller knees, indicating that the women knee joint pressure will be even higher. *This may be one explanation, among others, why gonarthrosis is more frequent among women than men (Danielsson & Hernborg 1970, Hernborg & Nilsson 1973).* The sex anatomical differences of the hip extensor moment arms are considerably smaller than that of the knee extensors, and therefore the hip force magnitudes in women and men are of the same magnitude (Németh & Ohlsén 1985). *This might mechanically explain why there is no sex difference in the prevalence of hip joint arthrosis but there is in the case of the knee (Danielsson 1966, Lindberg 1985).*

Using the biomechanical model the situation after patellectomy can be assessed for various knee flexion angles. An example of this is illustrated in Fig. 12 at 30 deg knee angle. The moment arm (dp) decreased 15 % which means that the knee extensor strength will decrease 15 % at this particular knee flexion angle. The patellar tendon angle (β) decreased from 21 to 5 deg, indicating that tibia will shear less in an anterior direction. The biomechanical effects of anteversion of the tuberositas tibiae can be assessed in a similar way.

During knee extension exercises at knee flexion angles straighter than 30–40 deg, there is a considerable anteriorly directed tibio-femoral shear force which gives rise to high forces in the anterior cruciate ligament (Lindahl & Movin 1967, Paulos et al. 1981, Arms et al. 1984, Noyes et al. 1984). Butler et al. (1980) showed that the anterior cruciate takes up 86 % of the anterior shear force. By means of the biomechanical model presented, such forces may be estimated and consequently, different adjust-

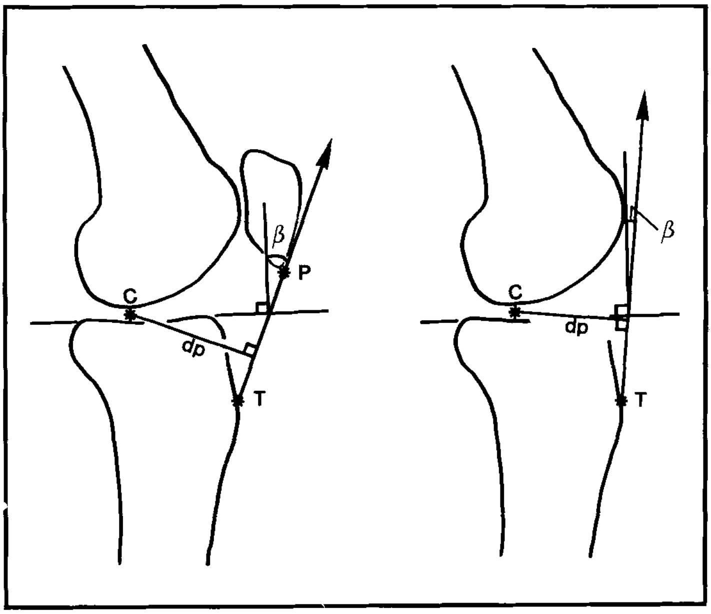

Fig. 12. Biomechanical effects of simulated patellectomy. Note the decrease of patellar tendon moment arm length (dp) and patellar tendon angle (β).

ments of knee extension exercises may be assessed with regard to *e.g.* anterior cruciate ligament stress. Such an example is given for isokinetic knee extension (Study VI), where it was concluded that having the resistance pad placed proximally on the anterior part of the lower leg considerably decreased the shear force and consequently the stress on the anterior cruciate ligament. Similar conclusions were found by Johnson (1982) when using a dual pad resistance and in a recent study by Jurist & Otis (1985). *Consequently, patients with repairs to or reconstructions of the anterior cruciate ligament should be advised to start their quadriceps exercise carefully at straighter knee angles and with the resistance force proximally on the lower leg.*

Over the last few degrees during knee extension against gravity (Fig. 9), the anteriorly directed shear force reached its maximum at the same time as the patello-femoral compressive force was low. The peak compressive force occurred at 50 deg knee flexion angle and now the shear force was low. However, during isokinetic knee extension, the shear force was also high at knee angles as flexed as 60 deg (Fig. 12b). Consequently, it is of interest to know and consider the internal force distribution during different quadriceps exercises in patients with different diagnoses such as patello-femoral arthrosis or anterior cruciate ligament deficiancy.

The force between the quadriceps tendon and femoral intercondylar groove (Fcq) was found to be considerable, especially when the knee was flexed to more than 90 deg (Fig. 10d). *It cannot be excluded that this force might compress the upper part of the suprapatellar bursa, and in some cases even cause pain.* This tendon-bone compression has been discussed earlier (Bishop 1977, Goodfellow et al. 1976, Goyman & Müller 1974, Grood et al. 1984), and Huberti & Hayes (1984) measured its magnitude using pressure sensitive film. However, its quantification in a biomechanical model has not earlier been performed.

A painful patello-femoral joint sensitive to high stresses (*e.g.* ostoarthrosis) is better exercised at straighter knee angles so that high flexing load moments and high compressive forces are avoided. Note that the force magnitudes received by the patello-femoral joint are about the same between 60 deg and 120 deg, so for patients with pain elicited from this joint there should be no reason to avoid 120 deg more than 60 deg. Exercises aiming at an optimal range of motion are of course also necessary, but should be undertaken mainly passively, *i.e.* with no or very little load. Patients with load-elicited pain from the patello-femoral joint might be advised to avoid knee angles above 30 deg under loaded conditions such as in deep squatting, lifting objects from the floor with flexed knees, ascending or descending stairs. The knee extensor apparatus is less activated at low knee angles and the patello-femoral compression is low. *A patello-femoral arthrosis patient might even be advised to lift with straight knees and flexed trunk* – if this does not cause low back pain – or to use technical aids.

Conclusions

The knee load moments and muscle activities during lifting and machine milking were quantified and a local knee biomechanical model presented. It was shown how to use the knee model in order to quantify the tibio-femoral and patello-femoral forces during ascending from the parallel squat in powerlifting and during isokinetic knee extension. The model may also be used to quantify knee joint forces during other knee extending activities and thus, to assess different work postures, training exercises and joint derangements.

The following main conclusions can be stated:

- Lifting a 12.8 kg box with flexed knees, from floor to table level, gave a maximum flexing knee load moment of about 50 Nm at the beginning of the lift, with a knee flexion angle of 90 deg. The quadriceps muscle was highly activated in order to counteract the load moment.
- No significant difference in knee joint load was seen with flexed knees during lifting the box from between or in front of the feet.
- Straight-knee lifts gave an extending knee load moment throughout the lift. During the final phase of all three lifts (SK, FKFF and FKC) there was an extending moment of about 55 Nm.
- During machine milking, the horizontal distance between the operator and the cow was found to be of considerable importance for the magnitude of the extending knee load moment in the straight-knee standing postures. The extending load moment decreased when the milker came closer to the cow and to the working area. Thus the horizontal distance to the cow should be considered when designing milking equipment.
- In flexed-knee postures, normalized EMG-activity in the quadriceps muscle was higher than that in the hamstrings muscles in straight-knee postures.
- The knee load moment in the standing flexed-knee postures was in the direction of flexion and comparatively low, while in the straight-knee postures it was in the direction of extension and comparatively high.

- In a static situation, with a defined hand-held implement and a horizontal distance to the working area, it is possible to determine what knee angle gives the least, or even zero, load moment on the knees. Flexing the knees gave a reduction of the extending load moment in the postures with small level differences between operator and cow. A knee angle of around 65 deg gave a zero load moment for implement weight and posture about the same as in Study II. This knee angle was not influenced by body weight.
- Women were found to have shorter patellar tendon moment arms than men, due to smaller knees, and thus develop about 20 % higher knee joint forces for the same knee extending moment.
- The angle between the tibial plateau and patellar tendon force changed considerably with knee flexion angle. Therefore, the anteroposterior tibio-femoral shear force was found to change direction from negative, with proximal tibia sliding posteriorly in relation to distal femur, to positive during the extension movement, indicating that high forces may arise in the anterior cruciate ligament when the knee is straight.
- The patellar tendon was found to be 25–30 % thinner and narrower than the quadriceps tendon, and the forces in these tendons at knee angles of 60–120 deg had a relation (Fp/Fq) of about 0.70–0.80. At straighter knee angles, the two forces were of the same magnitude.
- A prediction diagram was designed and used for quantifying patellar forces in women and men for given moment magnitude and knee angle.
- For a constant knee extending moment, the patello-femoral joint compressive force reached a maximum at 90 deg knee angle and decreased in magnitude beyond this knee angle.
- Above knee angles of 60 deg, the compressive force between the femoral intercondylar groove and the quadriceps tendon was quantified and found to increase linearly with increasing knee angle.
- The force magnitudes calculated using the knee joint biomechanical model developed, correlated well with forces measured experimentally by others, indicating that the model is valid.
- Performing a parallel squat while carrying a 382.5 kg weight induced a knee flexing load moment of 335–550 Nm. The *in vivo* force of a complete quadriceps tendon-muscle rupture was estimated to between 10,900 and 18,300 N.
- During isokinetic knee extension at slower speed (30 deg/sec), the patellar tendon force and tibio-femoral compressive force magnitudes peaked to almost 9 times BW and during the faster speed (180 deg/sec), to 5 times BW.
- During isokinetic knee extension at 30 deg/sec with the resistance pad distally, the anteriorly directed tibio-femoral shear force was found to be of considerable magnitude between 30 and 60 deg knee flexion angles. A proximal position of the resistance pad decreased the tibio-femoral shear force significantly, indicating that the anterior cruciate ligament receives lower stresses. This position of the pad was recommended for early postsurgical isokinetic rehabilitation of anterior cruciate ligament repairs and reconstructions.

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